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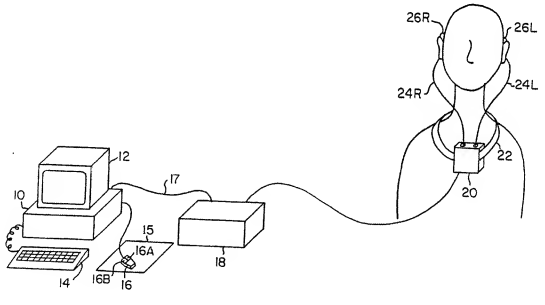
(54) Title: APPARATUS AND A METHOD FOR FITTING A HEARING AID



(57) Abstract

A hearing aid fitting system includes a computer having a graphical display which controls apparatus that can program and evaluate an adjustable, multi-band hearing aid. The fitting procedure implemented on this system determines the combined frequency response characteristic of the hearing aid and a client's ear, determines maximum amplification factors which can be implemented in each channel without inducing feedback, and automatically adjusts the hearing aid so that the combination of the client's loss curve, the determined combined frequency response characteristic, and the adjusted hearing aid match a target curve. The target curve and the loss curve may be graphically specified and adjusted using the computer.

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APPARATUS AND A METHOD FOR FITTING A HEARING AID

Technical Field

The present invention relates generally to the fitting of hearing aids and in particular to a computer controlled fitting procedure which allows the user to interact in the fitting process, using his particular hearing aid device.

Background Art

Generally, the correction of a hearing defect proceeds in several steps. First, a hearing test is performed. In this test, an audiometer is used to determine hearing thresholds for a range of distinct tones. In the next step, the test results are analyzed to classify the hearing defect. This classification may indicate that a particular type of hearing aid would be best for the person having the defect. The selected hearing aid is then adjusted to

have a frequency response characteristic that corrects the hearing defect indicated by the audiometric data. Finally, the hearing test is performed again, this time with the hearing aid in place, to determine the level of correction that has been achieved.

The process outlined above has several potential problems. First, there is an undesirable time delay between the audiometric testing of a client and the final fitting of the adjusted device. The length of this delay depends on the availability of a hearing aid that can correct the particular hearing defect, and the complexity of the required adjustments. Delays of between of one day and several weeks are not uncommon. These delays are not only a problem for the client who, obviously, would like to have his or her hearing aid as soon as possible, but delays are also a problem for the hearing aid provider. The provider's problems mainly relate to clients who may refuse to accept a device fitted several days before, claiming either that the testing procedure was improper or that the selection and adjustment of the device were improper.

Second, audiometric tests performed using asychronous receiver-transmitter (UART). instructions a standard audiometer do not adequately measure hearing loss and cannot account for the mechanical response of the ear with a hearing aid in place.

Consequently, the prescribed hearing aid may appear to perform well on the audiometric tests but may inadequately correct the users perception of other sounds such as speech. Moreover, a hearing aid that is fitted using only audiometric data cannot adequately compensate for resonances and nulls that result from mechanical interaction of the hearing aid and the ear canal.

Finally, in the steps described above, there is no opportunity for the individual being fitted to include his or her own preferences in the performance of the unit until after it has been selected and adjusted. To accommodate these preferences, the provider may have to make further adjustments to the device, causing further delays the fitting process.

Description of the Invention

The present invention is embodied apparatus and a method for fitting a person with a programmable hearing aid. A hearing aid fitted using this method desirably includes a plurality of separate channels each of which processes signals occupying a distinct band of frequencies. Data representing a client's hearing defect and data representing a target frequency response characteristic to correct the defect are entered into a computer. Responsive to

this data, the computer determines adjustment coefficients for the plurality of separate channels which most closely match the frequency response characteristic of the hearing aid to the target characteristic. These adjustment coefficients are then used to program the hearing aid while it is in place, coupled to the clients ear.

In accordance with a further application of this invention, before performing the steps set forth above, the receiver of a programmable hearing aid and a separate probe microphone are inserted into a client's ear canal. An exemplary probe microphone 410 is described below in reference to FIGURE 4. Using external signals applied to the hearing aid and response signals provided via the probe microphone, the combined frequency response characteristic of the hearing aid and the client's ear canal are determined. Next, using these same signals, a maximum amplification factor is determined for each of the plurality of channels by separately adjusting the level of amplification for each channel. Both the combined frequency response characteristic and the maximum amplification factors are entered into the computer, with the data representing the client's hearing defect and the target frequency response characteristic, to develop the plurality of coefficient values.

In accordance with another aspect of this invention, an operator may enter data regarding the client's hearing defect or data representing client preferences in the programming of the hearing aid into the computer directly on a graphic display of this data, using a pointing device.

Brief Description of the Figures

Figure 1 is an illustration of major elements of the fitting system.

Figures 2A and 2B are drawings which show the apparatus physically fitted to the client during the fitting operation.

Figure 3 is a block diagram of circuitry suitable for use as the interface unit of the system shown in Figure 1.

Figure 3A is a flow-chart diagram useful for explaining the operation of the interface unit shown in Figure 3.

Figure 4 is a block diagram of circuitry suitable for use as the chest pack of the system shown in Figure 1.

Figure 5 is a block diagram of an exemplary programmable hearing aid which may be fitted using the system shown in Figure 1.

Figure 6 is a schematic diagram, partially in block diagram form of an attenuator suitable for use in the hearing aid shown in Figure 5.

Figures 7, 8, 9 and 10 are flow-chart diagrams which illustrate the operation of the system shown in Figure 1.

Figure 11 is a diagram useful for explaining the operator interface to the system shown in Figure 1.

Detailed Description of the Figures

Figure 1 is a pictorial representation of the major elements of a hearing aid fitting system which includes an embodiment of the present invention. The system includes a computer 10 having a video display 12, a keyboard 14 and a pointer-type data entry device 16, commonly referred to as a "mouse." The computer 10 may be, for example, an IBM Personal Computer.™ An operator performing a fitting operation, enters data and commands for the fitting system into the computer 10 using the keyboard 14 and the mouse 16. The video

display 12 may be used to display audiometric data relating to the client being fitted.

In the present embodiment of the invention, a hearing aid is described which includes a 13 band equalizer. The frequency response characteristic of the hearing aid is determined by the net gain or amplification factors of each of the 13 frequency bands. All amplification factors and attenuation levels used in this application are in units of decibels (dB). Consequently, a positive attenuation level is equivalent to a negative amplification factor and the net gain of two cascaded amplifiers is determined by summing their individual gains.

The computer 10 is coupled by a data link 17 to an interface unit 18. The interface unit 18 is coupled to a chest pack 20, suspended from the neck of the client by a strap 22, and the chest pack 20 is coupled, by cables 24R and 24L, to one or two hearing aid devices 26R and 26L, respectively. Using the computer 10, interface unit 18 and chest pack 20, an operator can change the frequency response characteristic of the hearing aids 26R and 26L; apply audio signals through the hearing aids 26R and 26L to obtain audiometric data relating to the client; and, through probe microphones inserted into the client's ears, monitor the performance of the hearing aids 26.

Figures 2A and 2B are two views of a client's right ear which illustrate apparatus used in the fitting process. As shown in Figure 2A, the hearing aid device 26R is physically fitted behind the client's ear with an earmold 212 coupling the receiver of the hearing aid into the client's ear canal. A probe microphone, 210R is also inserted into the client's ear canal and is held in place by the earmold 212. An identical probe microphone 210L (not shown) is inserted into the client's left ear canal. A signal conducting path, 24A, for applying operating power and programming signals to the hearing aid 26R, is coupled via a connector (not shown) which may be located, for example, in the battery compartment (not shown) of the hearing aid 26R. In addition, a signal conducting path 24B, conveys signals from the probe microphone. The conduction paths 24A and 24B are physically coupled to form a single cable, 24R which couples the hearing aid 26R to the chest pack 20. The signals conveyed by these conduction paths are passed through the chest pack 20 as set forth below in reference to Figure 4, to or from the interface unit 18.

Circuitry suitable for use as the interface unit 18 is shown in Figure 3. Data communication between the interface unit 18 and the computer 10 is implemented by a receiver/transmitter unit 310. The unit 310 may be, for example, a conventional universal

asychronous receiver-transmitter (UART) circuit. Instructions and data from the computer 10 are passed through the unit 310 to a microprocessor 312. The microprocessor 312, using a program stored in a memory 314, processes the information received from the computer 10 to control a programmable signal generator 320, an analog-to-digital converter (ADC) 322 and a parallel-input serial-output register 316. The programmable signal generator used in this embodiment of the invention is a 5503 integrated circuit available from Ensonic Corporation. The bit-serial output signal of the register 316 is applied to the chest pack 20, via a synchronous input/output (I/O) device 318, to control the amplification factors for the 13 signal processing channels of the each of the hearing aids 26. The I/O device 318 also provides a synchronizing clock signal, CKPA and a signal, L/R, to the chest pack 20. The signal L/R selects either the hearing aid 26L or 26R to receive the programmed amplification factors.

Figure 3A is a flow-chart diagram of the program that controls the microprocessor 312. Functionally, this program accepts commands from the computer 10 and based on the type of command received, requests parametric data from the computer 10. The microprocessor 312 then uses the requested data to condition its various peripheral devices to perform the requested command. When the command has been

completed, the unit 18 sends a message to that effect, with any data that may have been requested, to the computer 10. As an error check, all transmissions between the computer 10 and interface unit 18 include a checksum. If an invalid checksum is detected, the receiving unit, 10 or 18, requests the originating unit to retransmit the data.

The first step, 350, in Figure 3A waits for a command from the computer 10. When a command is received, the microprocessor 312, at step 352 determines the command type. After processing any of the commands listed below, the microprocessor 312 branches back to the step 350 to await the next command.

If, at step 352, the command is a request to change the amplification factor or gain of a signal processing channel of one of the hearing aid devices 26R or 26L, the microprocessor 312, at step, 354, requests parametric information from the computer 10 to selects a hearing aid device 26L (left) or 26R (right), a channel (1 to 13) and a new attenuation level (0 to 63) for a programmable attenuator in the selected channel. The next step in the program, 355, sends a signal MUTE to the chest pack 20. This signal mutes the selected hearing aid while the attenuation level is changed.

Using the attenuator and gain selection data, the microprocessor 312, at step 356, loads 13 six-bit values representing the desired attenuation levels for the selected hearing aid 26R or 26L into the shift register 316. At step 358, the microprocessor 312 conditions the synchronous I/O unit 318 to apply the 13 six-bit values to the chest pack 20 as a bit-serial signal, PA. The I/O unit 318 also provides a clock signal CKPA to synchronize the chest pack 20 to the signal PA and a signal L/R to indicate whether the data is for the hearing aid 26L or 26R.

As an error check, the chest pack 20 automatically retransmits the data it receives to the interface unit 18. This data is received at step 360, via the synchronous I/O unit 318, which, in turn, provides the data to the microprocessor 312. The microprocessor 312 checks the received data against internally stored attenuation levels for the selected hearing aid. If an error is detected, the program branches back to step 358 and attempts to retransmit the data. If no error is detected, step 364 is executed which changes the state of the signal MUTE to reverse the muting operation performed in step 355 above. The effect if the signal MUTE on the hearing aids 26 is described below in reference to Figure 5.

When the final attenuation levels have been applied to the hearing aids 26, they may be stored

permanently into an electrically erasable read only memory (EEPROM) internal to the hearing aids 26. This command is conveyed to the chest pack 20 via a signal BURN provided by the microprocessor 18.

Another type of command which may be issued by the computer 10 is a request that a signal having a particular frequency and amplitude be sent to a selected one of the hearing aid devices 26. If a command of this type is detected at step 352, the microprocessor 312, at step 370, requests parameters from the computer 10 that select the left or right hearing aid and that define the frequency and amplitude of the signal to be sent. At step 372, the signal generator is conditioned to provide the requested signal. At step 374, a demultiplexer 321 is then conditioned by the microprocessor 312 to select a path to the right or left hearing aid 26R or 26L through the chest pack 20. As set forth below in reference to Figures 4 and 5, this audio signal is passed through the chest pack and applied internally to the selected hearing aid 26, bypassing the microphone input to the hearing aid. The parameters provided by the computer 10, through the microprocessor 312, may condition the signal generator 320 to provide different types of sound such as continuous tones, discontinuous tones or narrow band noise, and to provide sound signals having different durations.

The last type of command for this exemplary embodiment of the invention is a request to receive data from one of the probe microphones 210R or 210L, in the client's ear canals. The first step for this command, 380, obtains a parameter from the computer 10 which selects the probe microphone for the left or right hearing aid to provide the data. At step 382, the microprocessor 312 conditions a multiplexer 323 to select the signal provided by the selected hearing aid. The selected signal is applied to an RMS converter 324 which develops a direct current (DC) signal representing the sound pressure inside the client's ear canal. This signal is applied to the ADC 322 which, at step 384, digitizes the signal and applies it to the microprocessor 312. The microprocessor 312, at step 386, formats the data for transmission to the computer 10 at step 388.

Figure 4 is a block diagram illustrating circuitry suitable for use in the chest pack 20. In the chest pack 20, a shift register 412 is coupled to receive the signal PA provided by the interface unit 18. As set forth above, the signal PA is sent back to the interface unit 18 for error correction and detection immediately after it has been received. the signal conduction lines from

After the error checking procedure, the

stored values in the shift register 412 are applied to level shifting circuitry 414 which converts the signals from logic levels used by the interface unit 18 to lower logic levels used by the hearing aid circuitry. After this conversion, the signals are applied to the selected hearing aid 26R or 26L via a demultiplexer 416, as directed by the signal L/R provided by the interface unit 18.

The audio signals, SIGL and SIGR provided by the interface unit 18 are applied to the respective left and right hearing aid devices 26R and 26L. The signals 24A generated by the probe microphones 210 of the respective hearing aids 26R and 26L are amplified by the respective amplifiers 422 and 424, in the chest pack 20, to generate the signals MIC_R and MIC_L which are applied to the interface unit 18. On the hearing aid end of the chest pack 20, all of the signals for the left hearing aid 26L are grouped together to form the cable 24L and all of the signals for the right hearing aid are grouped together to form the cable 24R. As shown in Figure 1, these cables are applied to the respective hearing aid devices 26R and 26L. The signal MUTE and the signal CK_{PA} are passed through respective level shifting circuits 418 and 420 which adjust the logic levels produced by the microprocessor °[312 and I/O unit 318 to the +- 1.25 volt logic levels used in the hearing aids 26. The signal BURN is also passed through a level shifter 419, however, this

level shifter changes the signal BURN to a level sufficient to "burn" data into the EEPROMs of the hearing aids 26.

The chest pack 20 receives its operational power from the interface unit 18 and also provides operational power to the hearing aid devices 26R and 26L. This power supply circuitry is not shown to simplify the description of the invention.

Figure 5 is a block diagram circuitry suitable for use as either of the hearing aids 26R or 26L. This circuitry includes thirteen distinct signal processing channels which process signals in thirteen respective frequency bands. Each channel includes a band-pass filter and an programmable attenuator. The programmable attenuators may be adjusted, using digital interface circuitry, so that the hearing aid may be configured to compensate for a wide variety of hearing deficiencies. In addition, the five lowest frequency channels may be completely disabled in the presence of relatively high levels of ambient noise, both to save power by not processing this noise and to increase the intelligibility of words spoken over the ambient noise.

The hearing aid shown in Figure 1 includes a VLSI integrated circuit (IC) 500 which uses the power-efficient complementary metal oxide semiconductor

(CMOS) technology. Further power efficiency is gained by using switched-capacitor designs for the band-pass filters and programmable attenuators.

In the description set forth below, the circuitry used in this hearing aid is generally described in terms of its function. This general description is followed by a more detailed description of the programmable attenuators used in the various signal processing channels.

In Figure 5, sound waves are converted to electrical signals by a microphone 510, which applies the electrical signals to a compression amplifier 512. The amplifier 512 amplifies relatively quiet sounds and attenuates relatively loud sounds so that the signals provided by the amplifier remain substantially within a predetermined dynamic range. This dynamic range is compressed relative to the original dynamic range. The circuitry 512 amplifies signals in a frequency range from 100 Hz to 10 KHz. The output signals of the compressor amplifier 512 are applied to a further amplifier 514 which drives the 13 signal processing channels. The amplifiers 512 and 514 used in this embodiment of the invention provide a net amplification factor of 56 dB. These amplifiers are available as a single integrated circuit, the LD-512 manufactured by Gennum Corp.

As set forth above, each of the 13 signal processing channels includes a band pass filter and a programmable attenuator. The band-pass filters used in the respective signal processing channels are numbered 516 through 528 and their corresponding attenuators are numbered 540 through 552. The band-pass filters used in this embodiment of the invention are of conventional design. As shown in Figure 5, the band pass filter 516 has a one octave passband, the filters 517 and 518 have one-half octave passbands and the filters 520 through 528 have one-third octave passbands. The center frequencies for the filters 516 through 528 are 178 Hz, 338 Hz, 588 Hz, 776 Hz, 1.04 KHz, 1.44 KHz, 1.8 KHz, 2.3 KHz, 2.9 KHz, 3.6 KHz, 4.6 KHz, 5.8 KHz, and 7.2 KHz. The transition regions of the frequency response characteristics of these exemplary filters exhibit a roll-off of approximately 12 dB/octave. These frequency response characteristics were chosen so that the 13 channels convey approximately equal amounts of speech information. Circuitry suitable for use as one of these switched-capacitor band-pass filters is disclosed in U.S. Patent 4,622,440 entitled "Differential Hearing Aid With Programmable Frequency Response" which is hereby incorporated by reference.

Each of the thirteen band-pass filters passes signals in a respectively different band of frequencies. However, because each of the filters

passes signals outside of its passband at a reduced amplitude, there may be significant amounts of overlap in the signals passed by successive filters.

Each of the attenuators 540 through 552 is separately programmable via the chest pack 20 which provides programming signals, as set forth above, to a digital interface 92 and an electronically erasable programmable read-only memory (EEPROM) 94. Each of the attenuators 540 through 552 may be programmed to provide up to 40 dB of attenuation to the signals applied to its input terminal. An exemplary attenuator is described below in reference to Figure 6.

The output signals provided by the attenuators 540 through 552 are applied to summing and sample-and- hold circuitry 560. The circuitry 560 adds together all of the signals provided by the attenuators and samples the summed signal to remove any sampling artifacts introduced by the switched capacitor band- pass filters 516 through 528 and attenuators 540 through 552.

The output signal of the circuitry 560 is applied to an amplifier 570. The amplifier 570 is responsive to a volume control 571 to allow the user to adjust the level of sound produced by the hearing aid. During the fitting procedure, the volume control

571 is set to a predetermined position. The amplifier 570 provides an amplification factor between 0 dB and 40 dB. The signals produced by the amplifier 570 are applied to a final amplifier 572 which provides an amplification factor of 10 dB. In this embodiment of the invention, the final amplifier 572 is not a part of the IC 500. The output signals of the final amplifier 572 are applied to a final attenuator 574 which is a part of the IC 500.

The final attenuator 574 includes three resistors, RA, RB and RC and four switches, 574A, 574B, 574C and 574D. Each of these switches is controlled by a separate bit provided, in normal operation, by the EEPROM 594 or, during the fitting procedure by the chest pack 20 via the digital interface 592. When the switch 74D is closed, the output of the final amplifier 572 is coupled directly to the receiver 576, so there is no attenuation. If the switch 574D is open and any of the other switches 574A, 574B and 574C are closed, a resistance is inserted in the path between the final amplifier 572 and a receiver 576. The amount of attenuation provided by any combination of resistors depends on the relative impedances of the combined resistors and the receiver. The final attenuator 574 provides between 0 and 40 dB of attenuation. The amount of attenuation provided is adjustable via the computer 10, interface unit 18, chest pack 20, digital

interface 592 and EEPROM 594. This attenuator may be programmed in the same manner as the attenuators 540 through 552 except that the code for the different attenuation levels would be between 0 and 8 instead of between 0 and 63. The output signal of the final attenuator 574 drives the receiver 576 which produces sound waves in the ear of the user.

An output signal 513 of the compressor amplifier 512, which indicates the sound pressure level at the microphone 510, is applied to an ambient sound compensation circuit 515. The circuit 515 detects ambient sound levels having amplitudes in a predetermined range. Responsive to these detected levels, the circuit 515 selectively disables up to five of the lower frequency channels of the hearing aid circuitry using switches 529 through 533. This step successively disables the low frequency channels in the presence of high levels of ambient noise, both to save power and to increase the intelligibility of speech. Power is saved since the input signals to the band-pass filters and attenuators for the low frequency channels are disabled when high levels of signal are present. The intelligibility of speech is increased because the low frequency bands, which generally contain a large portion of the ambient noise, are deemphasized relative to higher frequency bands which contain a relatively large portion of the speech information.

Figure 6 is a schematic diagram, partially in block diagram form, of circuitry suitable for use as any of the programmable attenuators 540 through 552. This attenuator includes an operational amplifier 610 having a fixed capacitor 611 and a switched capacitor network 612 in a feedback loop from the output terminal, O, of the operational amplifier 610 to its inverting input terminal. In addition the attenuator includes six switched capacitor networks, 614, 616, 618, 620, 622 and 624 which are coupled between an input terminal, I, of the attenuator and the inverting input terminal of the operational amplifier 610. Each of the switched capacitor networks 614, 616, 618, 620, 622 and 624 includes a switch in its input path. These switches, 628, 630, 632, 634, 636 and 638 are controlled by signals G1, G2, G3, G4, G5 and G6, respectively, to selectively apply input signals provided via the input terminal I, through their respective switched capacitor networks, to the inverting input terminal of the operational amplifier 610.

The six signals G1 through G6 are the six-bit attenuation control signal provided by the digital interface 592. These signals are inverted by the respective inverters 640, 642, 644, 646, 648 and 650 to provide both inverted and noninverted versions of the signals to control the respective switches 628,

630, 632, 634, 636 and 638. Each of these switches and each switch in each of the switched capacitor networks may be, for example, a conventional CMOS transmission gate.

The switched capacitor networks 612, 614, 616, 618, 620, 622 and 624 are each responsive to two antiphasal clock signals, C1 and C2. These clock signals may have a frequency of, for example, 100 KHz and a duty cycle of slightly less than 50%. Except for the value of their capacitances, all of the switched capacitor networks are identical. The description of the structure and operation of the network 612, set forth below applies to the networks 614, 616, 618, 620, 622 and 624, which, for the sake of brevity, are not described in detail.

The switched capacitor network 612, includes four switches, S1, S2, S3 and S4 which alternately couple a capacitor 613 between the output terminal O and a source of reference potential (e.g. ground) on the one hand and the inverting input terminal of the operational amplifier 610 and ground on the other hand. The clock signal C1, which controls the switches S1 and S4, is substantially antiphasal to the signal C2 which controls the switches S2 and S3, so, when the switches S1 and S4 are open, the switches S2 and S3 are closed and vice-versa.

In a first time interval, when the switches S2 and S3 are closed, terminal A of the capacitor 613 is coupled to ground and terminal B is coupled to the output terminal O of the operational amplifier 610. During this time interval, the current provided by the amplifier 610 charges the capacitor 613 to a potential which depends on the amount of current provided by the amplifier 610, the capacitance of the capacitor 613 and the length of the first time interval. At the end of the first time interval, the switches S2 and S3 are opened and the switches S1 and S4 are closed coupling terminal A of the capacitor 613 to the inverting input of the amplifier 610 and terminal B to ground for a second time interval. In this configuration, the capacitor 613 is coupled to apply the inverse of the potential developed during the previous time interval to the inverting input of the amplifier 610. This potential is summed with potentials provided by the switched capacitor networks 614, 616, 618, 620, 622 and 624 to provide an input potential to the amplifier 610.

The fixed capacitor 611 configured in parallel with the switched capacitor network 612 acts to stabilize the amplifier 610 and to remove any high frequency artifacts of the switched capacitor processing. This capacitor conditions the circuitry shown in Figure 6 to operate as a low-pass filter having a cut-off frequency that is above the highest

audio frequency processed by the hearing aid but below the sampling frequency of the switched capacitor networks, that is to say, below the frequency of the clock signals C1 and C2. Thus, artifacts of the switched capacitor processing are rejected by the attenuator circuitry shown in Figure 6.

It is well known that switched capacitor networks such as 612, 614, 616, 618, 620, 622 and 624 may be modeled as resistors in analyzing the performance of the circuit. The value of the capacitor in a network is inversely proportional to the equivalent resistance value for the network. If each of the switched capacitor networks in the circuitry shown in Figure 6 were replaced by its equivalent resistance, the circuitry would be configured as a conventional class A amplifier, the gain of the amplifier being determined by the ratio of the feedback resistance to the input resistance.

In the circuitry shown in Figure 6, the values of the capacitors in the switched capacitor networks are chosen such that the gain of the amplifier is always fractional and so, the circuitry shown in Figure 6 always attenuates signals applied to its input terminal. The capacitors are selected such that the network 614 provides 20 dB of attenuation, and the networks 616, 618, 620, 622 and 624 divide up an additional 20 dB of attenuation into 0.5 dB steps.

Thus the attenuator shown in Figure 6 may be programmed to provide signal attenuation in a stepwise continuous range between approximately 0.5 dB and 40 dB.

The description set forth above relates to the apparatus used to fit the hearing aids 26 to a client. The actual procedure used in the fitting operation is described below. Since the apparatus described above is all controlled directly or indirectly by the computer 10, the fitting procedure is described in terms of a flow-chart diagram for a program implemented on the computer 10. In the present embodiment of the invention, this program is written in the C computer programming language. In addition, the program includes an interface to the Microsoft (tm) Windows program to provide the operator with a means for graphically entering commands and data.

Figure 7 is a flow-chart diagram which illustrates the overall procedure for fitting one hearing aid device. If a client is to be fitted with two hearing aids, each step shown in Figure 7 is performed twice, once for the right ear and once for the left.

The first step in Figure 7, step 710, calibrates the hearing aid device. This step, which

is described in greater detail with reference to Figure 8, determines the combined frequency response characteristic of the hearing aid and the client's ear canal. The next step, 712, finds the maximum amplification factor which may be realized in each of the 13 signal processing channels. The useful amplification factor for a particular channel or channels may be limited by positive feedback between the receiver 576 and microphone 710. The procedure for determining the maximum amplification factor is described below in reference to Figure 9.

In step 714, the hearing loss curve for the ear being fitted is specified. This step may be performed, for example, by having the operator use the computer 10 to plot a previously determined loss curve on a fitting work sheet as set forth below in reference to Figure 11. Alternatively, client's loss curve may be determined by performing standard audiometric tests, using the signal generator 320 of the interface unit 18 and the hearing aids 26 as an audiometer.

After the client's loss curve has been established, the operator, at step 716, specifies a target correction curve. This curve represents the desired corrected response for the client. The target curve may be entered manually on the fitting work sheet or determined automatically using one of several

standard formulas such as the Berger, POGO or N. A. L. formulas. The Berger formula is set forth in a paper by K. Berger et al. entitled "A Method of Hearing Aid Prescription", Hearing Instruments, July, 1978. The POGO formula is described in a paper by G. McCandless et al. entitled "Prescription of Gain/Output (POGO) for Hearing Aids", Hearing Instruments Vol. 34, No. 1, 1983. The N.A.L. formula is described in a publication by D. Byrne et al. entitled "The National Acoustic Laboratories' (NAL) New Procedure for Selecting the Gain and Frequency Response of a Hearing Aid". This publication is available from National Acoustic Laboratories, Chatswood, N.S.W., Australia.

In step 718, the computer 10 adjusts the amplification factors for each of the 13 signal processing channels to match the combined response, represented by the loss curve augmented by the hearing aid (i.e. the aided response curve), to the target curve. This step, which is described in detail below, with reference to Figure 10, calculates the best set of attenuation levels for the 13 signal processing channels of the hearing aid device and then applies these attenuation levels to the device. This procedure may not be able to exactly match the target curve due to limitations imposed by the combined frequency response characteristic of the hearing aid and the client's ear and by the maximum amplification factors that can be set in each channel as determined

at step 712.

After the hearing aid has been adjusted to implement the calculated amplification factors, a step 720 is executed to determine if further adjustment is desirable. An adjustment of this type may, for example, attempt to correct for differences between the actual performance of the hearing aid and its calculated performance, as determined by audiometric tests performed with the adjusted hearing aid in place. A further adjustment may be made, simply to accommodate a client's individual preferences, such as more high frequency or low frequency in the audio output signal. In response to these tests or preferences, the operator, at step 722, adjusts the target curve on the work sheet. The program then branches to step 718 to change the amplification factors of the 13 channels to match the new target curve. When, at step 720, it is determined that no further adjustment is desirable, the step 724 terminates the fitting procedure. This step may involve, for example, recording the client's name and address as well as the various response curves and amplification factors determined during the fitting procedure for future reference.

Figure 8 is a flow-chart diagram of the procedure used to calibrate one of the hearing aids 26R and 26L. The first step, 810, sets all of the

attenuators to a known attenuation level. This level may represent an attenuation level of 0 dB, 40 dB or any other value in-between. Next, at step 812, the variable CHANNEL is set to zero. At step 814, the variable CHANNEL is incremented to select the first channel. In the next step, 816, the computer 10 conditions the interface unit 18 to apply a sound signal to the selected hearing aid. This sound signal has an amplitude level that will not cause any discomfort to the client and has a frequency centered in the pass-band of the selected channel. It represents an acoustic signal having a known sound pressure level at the microphone input to the hearing aid.

While this signal is applied to the selected hearing aid, the computer 10, at step 818, requests the interface unit 18 to provide a measurement of the sound pressure in the client's ear canal, that is to say, a measure of the acoustic signal produced by the hearing aid in the client's ear. At step 820, the sound pressure level provided by the interface unit 18 is divided by the known input sound pressure level and the result is converted to an amplification factor in decibel units. The value produced by this operation represents the amplification factor of the combination of the hearing aid and the client`s ear canal at the selected frequency. This value is stored by the computer 10, step 822, for use in setting attenuation

levels for the 13 channels of the selected hearing aid. At step 824, the computer 10 branches back to step 814 if the selected channel is less than 13 and branches to step 826, ending the calibration process otherwise.

Figure 9 is a flow-chart diagram of the program that determines the maximum amplification factors that should be implemented in each of the 13 channels. This program is step 712 of Figure 7. If the amplification factor for a particular channel is too great, the acoustic signal provided by the receiver in the client's ear may be picked up at the microphone input of the hearing aid, causing the hearing aid to oscillate, producing a loud ringing noise. This noise is commonly known as feedback.

The process set forth in Figure 9 analyzes each channel separately and determines the largest amplification factor for the channel which will not produce feedback. This amplification factor is recorded by the computer 10 and used as an upper limit when the hearing aid is programmed to match the target curve. The interface unit 18 does not mute the selected hearing aid during this procedure, so that, referring to Figure 5, sound signals produced by the receiver 576 may be picked up by the microphone 510.

Step 910 of the program shown in Figure 9

sets the attenuators in all 13 channels to their maximum attenuation levels, that is to say, their minimum amplification factors. Step 912 initializes the variable CHANNEL and step 914 increments it, selecting the first channel. In step 916, the computer 10 conditions the interface unit 18 to reduce the attenuation level (i.e. increase the amplification factor) in the selected channel. Next, at step 918, the interface unit 18, under control of the computer 10, applies a narrow-band noise signal to the selected hearing aid. This signal has a very short duration, a frequency centered in the pass-band of the selected channel and an amplitude substantially equal to the maximum amplitude produced by the compressor amplifier 512 of Figure 5.

At step 920, the computer 10 conditions the interface unit 18 to monitor the signal received at the probe microphone after the narrow band noise signal has stopped. If a significant signal is measured at step 922, the hearing aid is assumed to be in an oscillatory state. In this instance, at step 924, the attenuation level is incremented by one for the selected channel to stop the feedback oscillation. The amplification factor corresponding to this incremented attenuation level is recorded at step 928 as the maximum amplification factor for the frequencies corresponding to the selected channel.

If at step 922, no significant signal is measured, then there is no feedback at the selected attenuation level. The step 926 determines if the current attenuation level for the selected channel is the lowest possible attenuation level. If so, the step 926 transfers control to the step 928 to mark the amplification factor corresponding to this attenuation level as the maximum amplification factor for the frequencies corresponding to the selected channel and to reset the attenuation level for the selected channel to its maximum value. Otherwise, the step 926 transfers control to step 916 to further reduce the attenuation level until either the minimum attenuation level is attained or feedback oscillation is detected.

Step 930, which is executed after step 928, checks the selected channel number against the maximum channel number, 13. If the selected channel is less than 13, control is transferred to step 914 to select the next channel. Otherwise, step 932 is executed ending the process.

When the maximum amplification factors for each of the 13 channels have been established, the operator specifies the client's loss curves and target curves and then attempts to correct the client's hearing loss by adjusting the frequency response characteristic of the hearing aid. This procedure is set forth above in reference to Figure 7.

Figure 10 is a flow-chart diagram of the program that adjusts the frequency response characteristic of a hearing aid 26R or 26L. This program corresponds to the step 718 of Figure 7. The steps 1010, 1012 and 1014 of Figure 10, generate a table of amplification factors for the hearing aids 26. The table has 13 columns representing the respective 13 signal processing channels, and the 64 rows representing the 64 respective attenuation levels for each channel. Each entry in the table represents an amplification factor provided by a channel of the hearing aid represented by the column number of the entry using an attenuation level represented by the row number.

In step 1010, a table of nominal amplification factors is derived. Each nominal amplification factor is generated from a common base factor. Referring to Figure 5, this base factor may be produced by summing fixed amplification factors for of the amplifiers 512, 514, 570 and 572 and then subtracting a predetermined fixed attenuation level for the variable attenuator 574. To generate the 64 nominal amplification factors for each of the 13 channels, each of the 64 variable attenuation levels for the channel are subtracted from the base amplification factor. This table may be generated, for example, by repeatedly invoking a FORTRAN

subroutine, included in Appendix A, once for each attenuation level in each channel.

In step 1012, the nominal amplification factors are combined with the calibration data produced by the algorithm set forth in Figure 8. In this step, the calibration response value, in decibels, for each channel is added to each of the 64 entries each row. In step 1014, entries in each column of the table representing amplification factors greater than the respective 13 maximum amplification factors determined by the algorithm set forth in Figure 9 are set to zero.

In steps 1016, 1018 and 1020 the computer 10 develops a first approximation of a frequency response characteristic which will correct the client's hearing loss to the response represented by the target curve. To generate this first approximation, the computer 10, at step 1016, determines the difference between the target curve and the loss curve at each of 13 frequencies, the center frequencies of the 13 signal processing channels. If, at step 1018, the difference value is determined to be less than the minimum or greater than the maximum amplification factor in the column, the respective minimum and maximum values are substituted for the difference value. At step 1020, the computer 10 searches each column of the table for the respective difference value determined in step

1016. The row number for each found entry represents the attenuation level applied to the respective channel.

The sum of the approximated frequency response characteristic defined by the attenuation levels provided by step 1020 and the loss curve is not a very good approximation of the target curve because the calculation of the amplification factor for any one channel of the hearing aid does not take into account contributions from the 12 other channels. To determine how good the approximation is, step 1022 computes the actual frequency response characteristic of the hearing aid by combining the contribution of all 13 channels for each of the 13 center frequencies. Two computer programs, written in the language FORTRAN, which implement this algorithm are included below in Appendices A and B.

The 13 values defining the frequency response characteristic of the hearing aid are added, at step 1023, to 13 corresponding values taken from the loss curve to produce an approximation of the corrected response.

The next step, 1024, subtracts 13 values of the target curve, representing the target response at the 13 center frequencies, respectively, from the 13 corrected response values produced in step 1023. This

operation produces a set of 13 error values which represent the difference between the approximate response curve and the target curve. In the step 1026, these error values are subtracted from error values produced in a prior iteration through the algorithm. For the first iteration, the prior error values are equal to zero.

If the difference between the present and prior error values is greater than a threshold, a step 1027 transfers control to step 1028. In step 1028, the computer 10 subtracts the present error values from the respective difference values provided by step 1016, above, to produce a new set of 13 difference values. Next, at step 1018, these difference values are limited and at step 1020, the amplification factor array is searched for the new difference values. This process continues through steps 1022, 1024 and 1026 and back to step 1018 until the step 1026 determines that the difference between the current and prior error values is less than the threshold. In this instance, step 1026 transfers control to step 1030 to end the algorithm.

Although, in the present embodiment of the invention, step 1026 only tests the current and prior error values against a threshold, it is contemplated that an additional test may be performed, one which branches to step 1030 if the number of iterations

through the filter adjusting algorithm is greater than a predetermined maximum value. This test would terminate the filter adjusting algorithm when the error values produced by step 1026 oscillates between two values, one representing an overshoot of the target curve and one representing an undershoot.

The material set forth above describes the apparatus used in the fitting procedure and the control of that apparatus through the computer 10. The following describes how the computer 10 is controlled by the operator. As described above, in reference to Figure 1, the operator uses a pointer-type input device, such as the mouse 16, to enter commands and data via the display device 12, using a graphic image such as that shown in Figure 11. In the present embodiment of the invention, this control mechanism is accomplished using a set of programs known as the Microsoft (R) Windows package, aviailable from Microsoft Inc. Using this package, a system designer may produce a graphic image, designate locations on the image as command entry points and regions of the image for data entry.

For the image shown in Figure 11, a control box 1112 is provided for entering commands to the system and a graph of frequency vs. hearing loss, 1110 is provided for entering and displaying data. The mouse 16 which is used to control the system is

represented on the by an arrow 1122 on the displayed image. When the operator moves the mouse 16 on the surface of a mouse pad 15, the arrow 1122 moves around the image in a corresponding manner.

The various boxes and circles in the control box 1112 are command points that allow the operator to establish and adjust a loss curve 1114 and a target curve 1118 for the client's left or right ear and to command the computer 10 to adjust the 13 channel attenuation factors of the selected hearing aid to generate an aided response curve 1120. In developing the aided response curve, the computer 10 makes use of a maximum gain curve, 1116 which is automatically generated during system initialization as set forth above in reference to Figure 9.

In general, the operator uses a command point by moving the mouse 16 on the mouse pad 15 to position the tip of the arrow 1122 over the desired command point. The command point is activated by pressing one of the mouse buttons 16A or 16B. When a command point allowing the adjustment or specification of one of the curves on the graph 1110 is activated, the mouse 16 may then be used to position the arrow on the graph 1110 to perform the change the curve.

The following example illustrates the use of the of the mouse 16 and the graphic display shown in

Figure 11 in the context of a typical fitting procedure. The first step in the procedure is to physically fit the client with left and/or right hearing aids 26 including earmolds 212 and to insert the probe microphones 210 into the client's ear canals. Next, the computer 10, interface unit 18 and chest pack 20 are initialized. During this step, the hearing aids 26 are calibrated to the client's ears and data for the maximum amplification factor curves 1116 are developed.

After initialization, a screen may be displayed that is similar to the one shown in Figure 11 except that the graph portion 1110 will have no curves. The operator then positions the arrow 1122 over the control point box 1130, marked "Right," and presses the button. In response to this selection, an "x" is displayed in the box 1130 and the maximum amplification factor 1116 for the right hearing aid 26R is displayed on the graph 1110.

Next, the operator specifies the loss curve. In this example, an air conduction (AC) loss curve was determined in a previous hearing test so the operator activates the box 1132 in the row marked "AC" and in the column marked "UnAided" and then activates the circle marked "S" in the same row. With these command points activated, the operator moves the arrow 1122 to the graph 1110, positioning it successively over the

points 1114A through 1114E and pressing one of the mouse buttons 16A or 16B to establish points for the curve. When all of the points have been specified, the computer 10 interpolates interstitial points to generate the displayed curve 1114. By activating the corresponding command points in the row marked AC(M) audiometric data developed for one ear using air conduction signals with masking noise applied to the other ear may be specified in addition to or in place of the air conduction data.

Alternatively, the loss curve may be determined by an air conduction or air conduction with masking audiometric testing procedure (not shown) using the hearing aids 26 to provide sound signals to the client's ears. At the conclusion of this audiometric testing, the loss curve 1114 would be stored by the computer 10. When the control point 1132 is specified, this curve would automatically appear on the graph 1110.

The next step in the fitting procedure is to generate the target correction curve 1118. This step is performed by activating the box 1136 marked "Target", then repeatedly pressing the mouse button while the arrow 1122 is positioned over the box 1138 until the desired calculation method appears in the box 1140. Finally, the arrow is positioned over the circle 1142 and the mouse button is pressed to

activate the circle 1142, conditioning the computer 10 to calculate data for the target curve 1118.

The last step in the procedure is to calculate the aided loss curve 1120. This curve represents the best match to the target curve 1118 which can be achieved adjusting the hearing aid 26R. This process is initiated by activating the box 1143. The result of this process is the curve 1120.

After performing the initial fit operation, the hearing aid 26R may be further adjusted by using the mouse to change the shape of the target curve 1118 and then activating the box 1143 to attempt to fit the client's corrected hearing curve 1120 to the adjusted target curve. The target curve is adjusted in several steps. First, the command point 1136 is activated, then the arrow is moved so that its tip covers a point on the curve 1118. This point may be moved vertically on the graph 1110 by pressing a mouse button - at which point the arrow becomes a cross-hair - and, while the button is pressed, moving the mouse to move the cross-hair to the desired position. When the mouse button is released, the curve 1118 is redrawn to pass through each point of the adjusted target curve including the point that has just been moved.

The new curve, which passes through all of the specified points, is calculated as follows.

Coordinates representing the first three specified points for the curve are read from the display. These coordinates are applied to a polynomial fitting algorithm which attempts to find a polynomial of degree three or less which passes through the specified points. However, only the portion of this polynomial that passes between the first two points is used. This process is repeated using the previous second and third points as new first and second points, respectively, and using the next specified point on the curve 1118 as the new third point. When the final three specified points are evaluated, the curve generated by the polynomial fit algorithm is used for the segment of the response curve 1118 spanning all three points. This same algorithm may be used to calculate any of the curves which may be specified using the mouse 16 as set forth above.

At any point when a corrected loss curve 1120 is available, the hearing aid 26R may be temporarily programmed to realize that response. The response may then be evaluated by the client and may be adjusted, as set forth above, to accommodate his preferences. When the final adjustments have been made, they may be permanently programmed into the nonvolatile EEPROM 594 of the hearing aid. However, since the EEPROM is electronically erasable, the same hearing aid may be re-adjusted at a later date to correct errors in the original fitting procedure or to

track a deteriorating condition in the client.

Although the present invention has been described in terms of an exemplary embodiment, it is contemplated that it may be practiced as outlined above with modifications within the spirit and scope of the appended claims.

APPENDIX A Page 1

```
C     THIS PROGRAM CALCULATES THE MAGNITUDE AND PHASE ANGLE
C     OF A SINGLE BAND-PASS FILTER
C
C     THE GAIN CAN BE SET BY SCALING THE MAGNITUDE OF THE
C     OUTPUT
C
C
      REAL*4 FREQ,GAIN,RE,IM,PHASE,FREQ1,FREQ2,DB
      REAL*8 WT,W,T,FQ,POWER,CK0,CK1,CK2,CK3,C3,C4,K0,K1,
     *K2,K3
      INTEGER K,L,LQ
      COMPLEX*8 O
      COMPLEX*16,Z,V,X,Y,H,S
      WRITE(*,*)" ENTER CAP VALUES... "
      WRITE(*,*)"     CKO "
      READ(*,*)CKO
      WRITE(*,*)" CK1 AND CK2 "
      READ(*,*)CK1
      WRITE(*,*)"     CK3"
      READ(*,*)CK3
      WRITE(*,*)"     C3 AND C4"
      READ(*,*)C3
      FQ=32000
      T=1/FQ
      CK0=CK0*1E-12
      CK1=CK1*1E-12
      CK2=CK1
      CK3=CK3*1E-12
      C3=C3*1E-12
      C4=C3
      K0=CK0/C3
      K1=CK1/C3
      K3=CK3/C3
      K2=CK2/C4
      OPEN(3,FILE="HOUT.ALL",STATUS="NEW")
C
C     BEGIN LOOP
C
C     RANGE IS 128HZ TO 16.4KHZ
C
C     H(Z)=[CKO/(1+K3)](Z-1)>/[Z**2-Z(2+K3-K1*K2)/
C     (1+K3)+1/(1+K3)]          FIG.5.5-1 (b)    Sanchez
C
```

APPENDIX A Page 2

```
C
      POWER=6.95
      DO 20 I=1,141
      POWER=POWER+.05
      W=2**POWER
      WT=W*T=6.28
      V=DCMPLX(DBLE(0.),WT)
      Z=CDEXP(V)
C     CALCULATE H(Z)
      X=(Z**2)-Z*(2+K3-K1*K2)/(1+K3)+1/(1+K3)
      Y=KO*(Z-1)/(X*(1+K3))
      O=Y
      IM=AIMAG(O)
      RE=REAL(O)
C
C     CALCULATE GAIN AND PHASE
C
      GAIN=CDABS(Y)
      PHASE=ATAN2(IM,RE)
      DB=ALOG10(GAIN)
      DB=DB*20
C     WRITE(*,*) W,GAIN,PHASE
      WRITE(3,200) INT(W),GAIN
      WRITE(5,200) INT(W),PHASE
      WRITE(4,201) INT(W),GAIN,PHASE
      WRITE(6,200) INT(W),DB
200   FORMAT(1X,I5,10X,F8.3)
201   FORMAT(1X,I5,10X,F8.3,10X,F8.3)
202   FORMAT(1X,F7.2)
20    CONTINUE
      CK0=CKO*1E12
      CK1=CK1*1E12
      CK2=CK2*1E12
      CK3=CK3*1E12
      C3=C3*1E12
      C4=C4*1E12
      WRITE(3,202)CKO
      WRITE(3,202)CK1
      WRITE(3,202)CK2
      WRITE(3,202)CK3
      WRITE(3,202)C3
      WRITE(3,202)C4
      STOP
      END
```

APPENDIX B

```
C     THIS PROGRAM SUMS TOGETHER VARIOUS FREQUENCY RESPONSES
C     PREVIOUSLY CALCULATED.  THIS IS TO OBSERVE
C     HOW THE PHASE MAGNITUDE OF VARIOUS PASSBANDS WILL
C     AFFECT EACH OTHER.  ANY COMBINATION OF PREVIOUSLY
C     CALCULATED FREQUENCY RESPONSES CAN BE USED, AS THE
C     PROGRAM PROMPTS THE USER FOR THE NUMBER OF INPUTS
C     AND THE NAME OF EACH.
C
      REAL*4 GAIN(20,141),RE,IM,PHASE(20,141)DPHASE
      REAL*8 WT,W,T,DGAIN,DB
      INTEGER K,L,LQ,FREQ(141)
      COMPLEX*8 O
      COMPLEX*16 Z,V,X,Y,H,S
      CHARACTER*10 IT
      WRITE(*,*)" HOW MANY RESPONSES ARE YOU SUMMING?"
      READ(*,*)K
      OPEN(4,FILE="S.ALL",STATUS="NEW")
      DO 40 I=1,K
      WRITE(*,*)"WHAT INPUT FILE?"
      READ(*,"(A6)")IT
      OPEN(3,FILE=IT,STATUS="OLD")
      DO 45 L=1,141
      READ(3,*)FREQ(L),GAIN(I,1),PHASE(I,L)
C     WRITE(*,*)FREQ(L),GAIN(I,L),PHASE(I,L)
45    CONTINUE
40    CONTINUE
C     DO THE ADDITION
      DO 60 L=1,141
      RE=0
      IM-0
      DO 55 I=1,K
      RE=RE+GAIN(I,L)*COS(PHASE(I,L))
      IM=IM+GAIN(I,L)*SIN(PHASE(I,L))
55    CONTINUE
      V=DCMPLX(RE,IM)
      DGAIN=COABS(V)
      DPHASE=ATAN2(IM,RE)
      DB=ALOG10(DGAIN)
      DB=DB*20
      WRITE(4,201) FREQ(L),DGAIN,DPHASE
60    CONTINUE
200   FORMAT(1X,I3,10X,F8.3)
201   FORMAT(1X,I5,10X,F.3,10X,F8.3)
      STOP
      END
```

We Claim:

1. In a system for fitting a hearing aid having an adjustable frequency response characteristic and having a receiver for amplifying sounds in the ear canal of a client to compensate for a hearing loss, a method for adjusting the frequency response characteristic of said hearing aid comprising the steps of:

inserting the receiver of said hearing aid into the ear canal of the client;

appla signal, having a predetermined amplitude to an input port of said hearing aid;

measuring the level of the sound produced in the ear canal of the client in response to said signal;

calculating, from the applied signal and the measured level of sound produced in the ear canal, a mechanical frequency response characteristic representing the interaction of the hearing aid and the ear canal;

calculating, from the mechanical frequency response characteristic and from the hearing loss, a frequency response characteristic for said hearing aid

to compensate therefor.

2. The method set forth in Claim 1 wherein the step of measuring the level of sound produced in the ear canal of the client includes the steps of:

inserting a probe microphone into the ear canal of the client; and

calculating the root-mean-square of the signal produced by the microphone in response to the signal applied to the hearing aid.

3. In a system for fitting a hearing aid having an adjustable amplification factor and having a receiver for providing amplified sounds in the ear canal of a client to compensate for a known hearing loss, a method for adjusting the amplification factor of said hearing aid comprising the steps of:

a) inserting the receiver of said hearing aid into the ear canal of the client;

b) adjusting the amplification factor of said hearing aid to have a predetermined level

c) applying an audio frequency signal,

having a predetermined amplitude, to an input port of said hearing aid;

d) measuring the level of the amplified sound produced in the ear canal of the client in response to said audio frequency signal;

e) calculating, from the applied signal and the measured level of sound produced in the ear canal, a mechanical frequency response characteristic representing the interaction of the hearing aid and the ear canal;

f) calculating, from the mechanical frequency response characteristic and from the known hearing loss, a final amplification factor for said hearing aid to compensate for said known hearing loss;

g) adjusting the amplification factor of said hearing aid to equal said final amplification factor.

4. The method set forth in Claim 3 wherein:

step A further includes the step of inserting a probe microphone into the ear canal of the client; and

step D includes the step of generating, from the signal provided by said probe microphone, an indication of the sound pressure level produced in the ear canal of the client in response to said signal.

5. The method set forth in Claim 3 wherein step F includes a method for determining a maximum amplification factor which may be used by said hearing aid, said method comprising the steps of:

adjusting the amplification factor of said hearing aid to a predetermined minimum value;

applying a further audio frequency signal to said hearing aid during a predetermined time interval;

measuring the sound level in the ear canal of the client after said predetermined time interval and setting an indicator variable in response to said measured sound level being greater than a threshold value;

increasing the amplification factor of said hearing aid in response to said indicator variable not being set;

decreasing the amplification factor of said hearing aid in response to said indicator variable

being set; and

storing said decreased amplification factor as said maximum amplification factor.

6. The method set forth in Claim 3 wherein step F includes the steps of:

specifying a target frequency response characteristic representing a desired level of hearing for the client fitted with the hearing aid;

calculating said final amplification factor, such that the combination of said final amplification factor, said known hearing loss and said mechanical frequency response characteristic tends to match said target frequency response characteristic.

7. The method set forth in Claim 6 wherein the step of specifying said target frequency response characteristic includes the steps of:

successively positioning a marker on a two-dimensional graphic display to first, second and third locations, representing respective first, second and third points in a graphic representation of said target frequency response characteristic;

entering coordinates representing said first, second and third locations in a table;

calculating from the coordinates representing said first, second and third locations, a portion of said target frequency response characteristic spanning said first and second points.

8. In a system for fitting a hearing aid having an adjustable amplification factor having a receiver for amplifying sounds in the ear canal of a client to compensate for a hearing loss, a method of determining a maximum value for said adjustable amplification factor, comprising the steps of:

a) adjusting the amplification factor of said hearing aid to a predetermined value;

b) applying an audio frequency signal to said hearing aid during a predetermined time interval;

c) measuring the sound level in the ear canal of the client after said predetermined time interval and setting an indicator variable in response to said measured sound level being greater than a threshold value;

d) increasing the amplification factor of

said hearing aid in response to said indicator variable not being set;

e) decreasing the amplification factor of said hearing aid in response to said indicator variable being set;

f) storing said resultant amplification factor as said maximum value.

9. The method set forth in Claim 8 wherein:

step D further includes the step of repeating steps B through D while said amplification factor is less than a predetermined maximum possible value and until said indicator variable is set after step C; and

step E further includes the steps of resetting said indicator variable and repeating steps B, through E until said indicator variable is not set after step C.

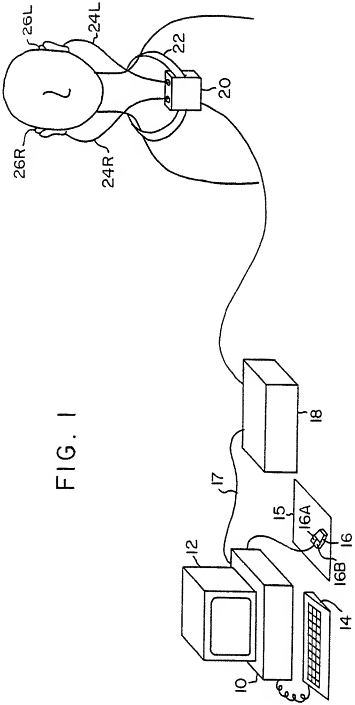


FIG. 1

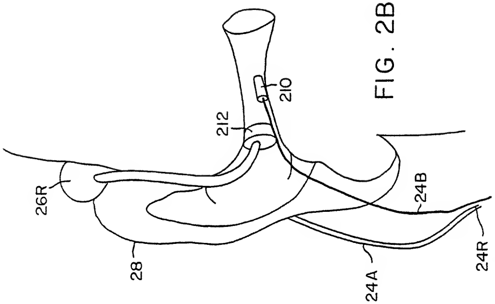


FIG. 2B

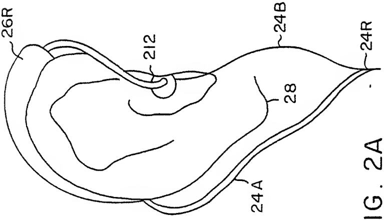
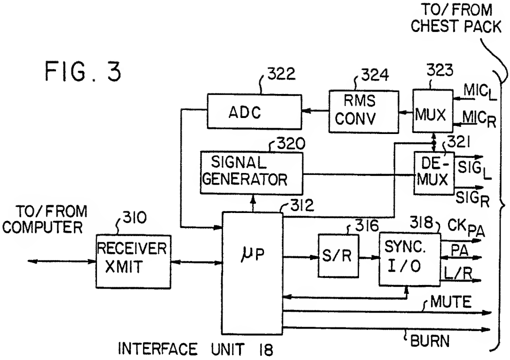


FIG. 2A

FIG. 3


TO/FROM CHEST PACK
322
324
323
ADC
RMS CONV
MUX
MIC_L
MIC_R
320
321
SIGNAL GENERATOR
DE-MUX
SIG_L
SIG_R
TO/FROM COMPUTER
310
312
RECEIVER XMIT
μP
316
318
CK_{PA}
S/R
SYNC. I/O
PA
L/R
MUTE
BURN
INTERFACE UNIT 18

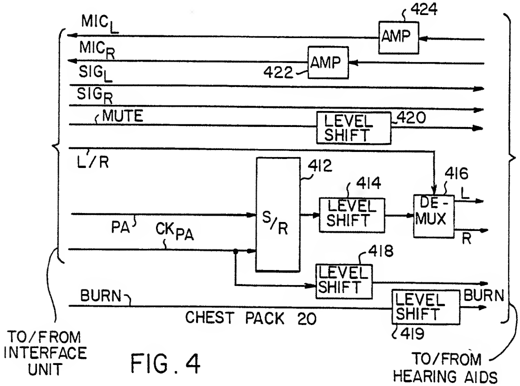

424
MIC_L
AMP
MIC_R
AMP
SIG_L
422
SIG_R
MUTE
LEVEL SHIFT
420
L/R
412
416
414
L
LEVEL SHIFT
DE-MUX
S/R
PA
CK_{PA}
R
418
LEVEL SHIFT
BURN
LEVEL SHIFT
BURN
CHEST PACK 20
419
TO/FROM INTERFACE UNIT
TO/FROM HEARING AIDS

FIG. 4

FIG. 3A

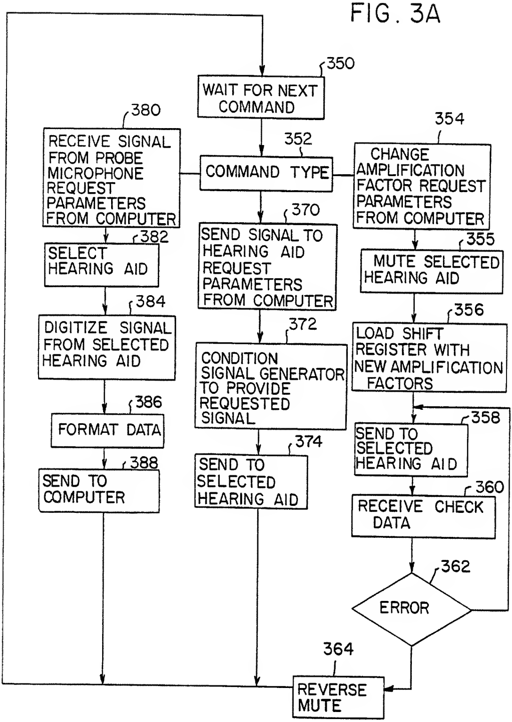

350
WAIT FOR NEXT COMMAND
352
COMMAND TYPE
380
RECEIVE SIGNAL FROM PROBE MICROPHONE REQUEST PARAMETERS FROM COMPUTER
382
SELECT HEARING AID
384
DIGITIZE SIGNAL FROM SELECTED HEARING AID
386
FORMAT DATA
388
SEND TO COMPUTER
370
SEND SIGNAL TO HEARING AID REQUEST PARAMETERS FROM COMPUTER
372
CONDITION SIGNAL GENERATOR TO PROVIDE REQUESTED SIGNAL
374
SEND TO SELECTED HEARING AID
354
CHANGE AMPLIFICATION FACTOR REQUEST PARAMETERS FROM COMPUTER
355
MUTE SELECTED HEARING AID
356
LOAD SHIFT REGISTER WITH NEW AMPLIFICATION FACTORS
358
SEND TO SELECTED HEARING AID
360
RECEIVE CHECK DATA
362
ERROR
364
REVERSE MUTE

FIG. 5-1

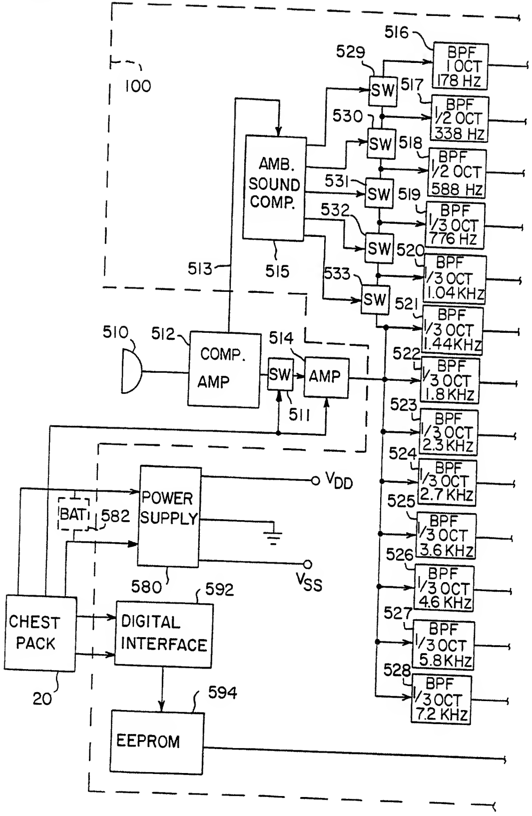
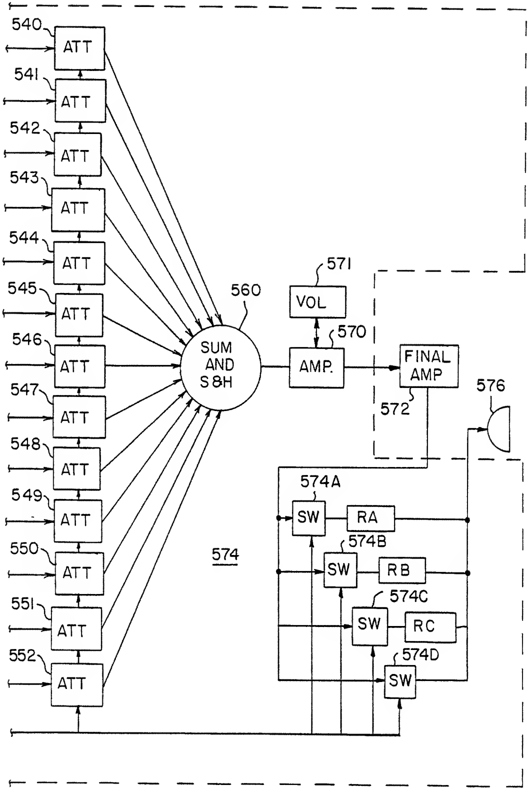

100
516
529
BPF 1 OCT 178 Hz
517
SW
530
BPF 1/2 OCT 338 Hz
518
SW
BPF 1/2 OCT 588 Hz
AMB. SOUND COMP.
531
SW
519
BPF 1/3 OCT 776 Hz
532
SW
520
BPF 1/3 OCT 1.04 KHz
513
515
533
SW
521
BPF 1/3 OCT 1.44 KHz
510
512
COMP. AMP
514
SW
AMP
511
522
BPF 1/3 OCT 1.8 KHz
523
BPF 1/3 OCT 2.3 KHz
POWER SUPPLY
V_{DD}
BAT
582
524
BPF 1/3 OCT 2.7 KHz
525
BPF 1/3 OCT 3.6 KHz
526
BPF 1/3 OCT 4.6 KHz
580
592
V_{SS}
CHEST PACK
DIGITAL INTERFACE
527
BPF 1/3 OCT 5.8 KHz
528
BPF 1/3 OCT 7.2 KHz
20
594
EEPROM

FIG. 5-2


540
541
542
543
544
545
546
547
548
549
550
551
552
ATT
560
SUM AND S&H
571
VOL
570
AMP.
FINAL AMP
572
576
574A
574B
574C
574D
SW
RA
RB
RC
574

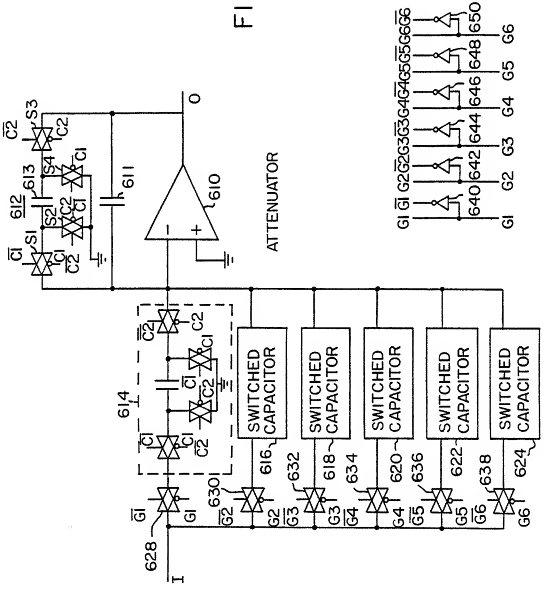

ATTENUATOR
SWITCHED CAPACITOR
SWITCHED CAPACITOR
SWITCHED CAPACITOR
SWITCHED CAPACITOR
SWITCHED CAPACITOR
610
611
612
613
614
616
618
620
622
624
628
630
632
634
636
638
640
642
644
646
648
650
S1
S2
S3
S4
I
O

FIG. 6

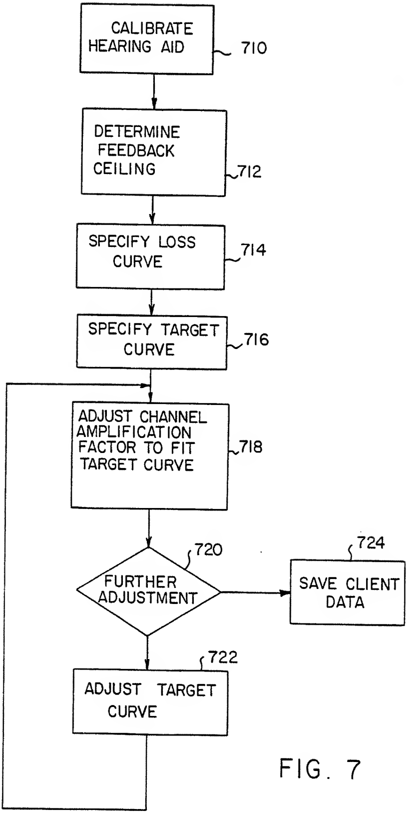

CALIBRATE HEARING AID
710
DETERMINE FEEDBACK CEILING
712
SPECIFY LOSS CURVE
714
SPECIFY TARGET CURVE
716
ADJUST CHANNEL AMPLIFICATION FACTOR TO FIT TARGET CURVE
718
720
FURTHER ADJUSTMENT
724
SAVE CLIENT DATA
722
ADJUST TARGET CURVE

FIG. 7

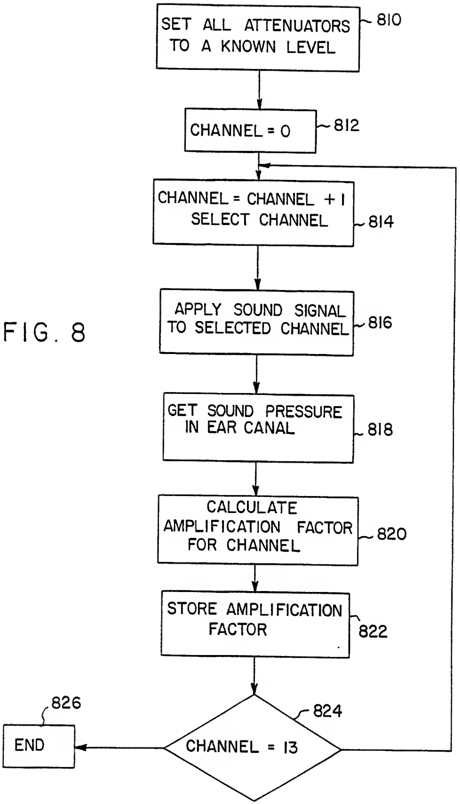

SET ALL ATTENUATORS TO A KNOWN LEVEL
810
CHANNEL = 0
812
CHANNEL = CHANNEL + I SELECT CHANNEL
814
APPLY SOUND SIGNAL TO SELECTED CHANNEL
816
GET SOUND PRESSURE IN EAR CANAL
818
CALCULATE AMPLIFICATION FACTOR FOR CHANNEL
820
STORE AMPLIFICATION FACTOR
822
CHANNEL = 13
824
END
826

FIG. 8

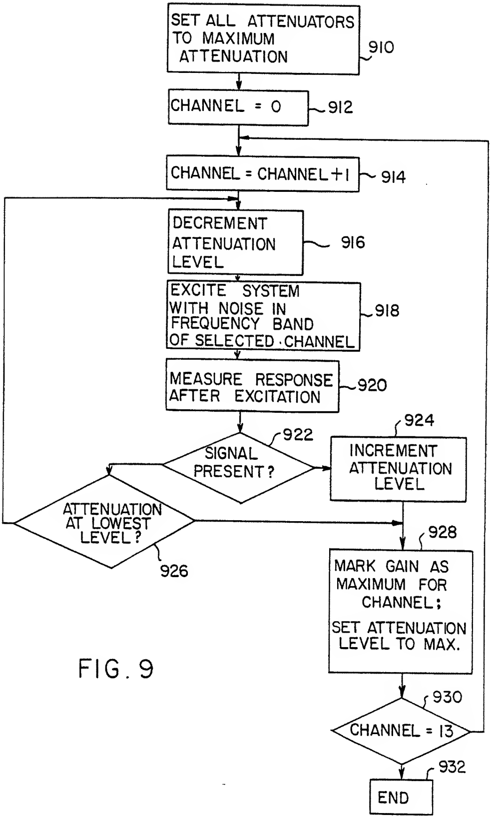

SET ALL ATTENUATORS TO MAXIMUM ATTENUATION
910
CHANNEL = 0
912
CHANNEL = CHANNEL + 1
914
DECREMENT ATTENUATION LEVEL
916
EXCITE SYSTEM WITH NOISE IN FREQUENCY BAND OF SELECTED CHANNEL
918
MEASURE RESPONSE AFTER EXCITATION
920
922
SIGNAL PRESENT?
924
INCREMENT ATTENUATION LEVEL
ATTENUATION AT LOWEST LEVEL?
926
928
MARK GAIN AS MAXIMUM FOR CHANNEL;
SET ATTENUATION LEVEL TO MAX.
930
CHANNEL = 13
932
END

FIG. 9

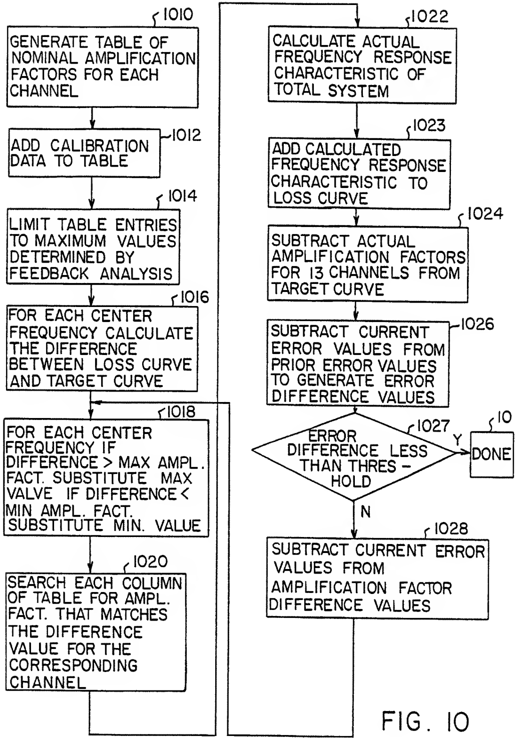
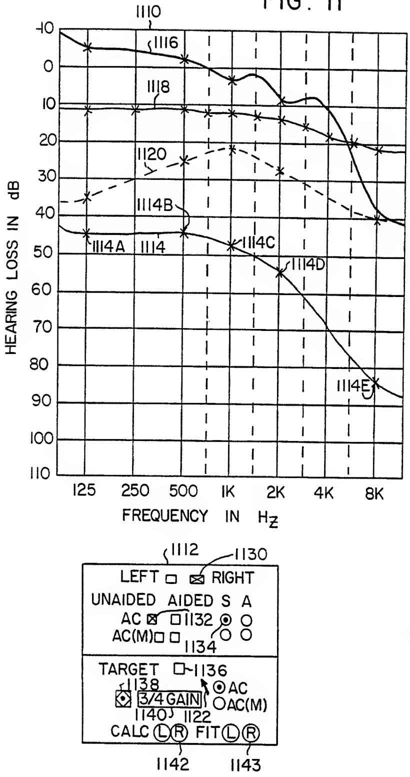

1010
GENERATE TABLE OF NOMINAL AMPLIFICATION FACTORS FOR EACH CHANNEL
1012
ADD CALIBRATION DATA TO TABLE
1014
LIMIT TABLE ENTRIES TO MAXIMUM VALUES DETERMINED BY FEEDBACK ANALYSIS
1016
FOR EACH CENTER FREQUENCY CALCULATE THE DIFFERENCE BETWEEN LOSS CURVE AND TARGET CURVE
1018
FOR EACH CENTER FREQUENCY IF DIFFERENCE > MAX AMPL. FACT. SUBSTITUTE MAX VALVE IF DIFFERENCE < MIN AMPL. FACT. SUBSTITUTE MIN. VALUE
1020
SEARCH EACH COLUMN OF TABLE FOR AMPL. FACT. THAT MATCHES THE DIFFERENCE VALUE FOR THE CORRESPONDING CHANNEL
1022
CALCULATE ACTUAL FREQUENCY RESPONSE CHARACTERISTIC OF TOTAL SYSTEM
1023
ADD CALCULATED FREQUENCY RESPONSE CHARACTERISTIC TO LOSS CURVE
1024
SUBTRACT ACTUAL AMPLIFICATION FACTORS FOR 13 CHANNELS FROM TARGET CURVE
1026
SUBTRACT CURRENT ERROR VALUES FROM PRIOR ERROR VALUES TO GENERATE ERROR DIFFERENCE VALUES
1027
ERROR DIFFERENCE LESS THAN THRES - HOLD
Y
N
10
DONE
1028
SUBTRACT CURRENT ERROR VALUES FROM AMPLIFICATION FACTOR DIFFERENCE VALUES

FIG. 10

FIG. 11


1110
1116
1118
1120
1114B
1114A
1114
1114C
1114D
1114E
-10
0
10
20
30
40
50
60
70
80
90
100
110
HEARING LOSS IN dB
125
250
500
1K
2K
4K
8K
FREQUENCY IN Hz
1112
1130
LEFT
RIGHT
UNAIDED AIDED S A
AC
1132
AC(M)
1134
TARGET
1136
1138
3/4 GAIN
AC
AC(M)
1140
1122
CALC L R FIT L R
1142
1143

INTERNATIONAL SEARCH REPORT

International Application No PCT/US 90/01175

I. CLASSIFICATION OF SUBJECT MATTER (if several classification symbols apply, indicate all)[6]

According to International Patent Classification (IPC) or to both National Classification and IPC

IPC5: A 61 B 5/12, H 04 R 25/00

II. FIELDS SEARCHED

Minimum Documentation Searched[7]

Classification System	Classification Symbols
IPC5	H 04 R, A 61 B

Documentation Searched other than Minimum Documentation to the Extent that such Documents are Included in Fields Searched[8]

III. DOCUMENTS CONSIDERED TO BE RELEVANT[9]

Category *	Citation of Document,[11] with indication, where appropriate, of the relevant passages[12]	Relevant to Claim No.[13]
A	US, A, 4596902 (S. GILMAN) 24 June 1986, see column 2, line 29 - line 68; abstract --	1-9
A	US, A, 4289143 (F. CANAVESIO ET AL) 15 September 1981, see the whole document --	1-9
A	DE, C2, 3113849 (SIEMENS AG) 11 September 1986, see the whole document -- --------	1-9

* Special categories of cited documents: [10]

"A" document defining the general state of the art which is not considered to be of particular relevance

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"Y" document of particular relevance, the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art.

"&" document member of the same patent family

IV. CERTIFICATION

Date of the Actual Completion of the International Search	Date of Mailing of this International Search Report
28th May 1990	21. 06. 90
International Searching Authority	**Signature of Authorized Officer**
EUROPEAN PATENT OFFICE	M. Peis M. PEIS

ANNEX TO THE INTERNATIONAL SEARCH REPORT
ON INTERNATIONAL PATENT APPLICATION NO.PCT/US 90/01175

SA 35187

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The members are as contained in the European Patent Office EDP file on 07/05/90
The European Patent office is in no way liable for theseparticulars which are merely given for the purpose of information.

Patent document cited in search report	Publication date	Patent family member(s)	Publication date
US-A- 4596902	24/06/86	NONE	
US-A- 4289143	15/09/81	EP-A-B- 0014324	20/08/80
DE-C2- 3113849	11/09/86	NONE	

For more details about this annex : see Official Journal of the European patent Office, No. 12/82

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